Abstract: A superconducting field-effect transistor (SuFET) based transducer (sensor) with carbon nanotubes (CNT) or pickup coil kind of input circuit for the nerve and neuron impulses, DNA recombination signals, flows of biochemical molecules, micro- and nanoscopy, and biosusceptibility has been designed. A nanoSuFET with a high-temperature superconducting channel is introduced into the nerve fibre or brain tissue for transducing their signals in both directions. Pickup coils are implanted into an organism in order to obtain the natural or artificially excited biosignals from the organs and tissues. The range of picked up signals varies from 0.6 nA to 10 µA with frequencies from 20 to 2000 Hz. The output signal lies in the range of \((5\pm5)V, (7\pm0) \cdot 10^{7}/\text{cm}^2\) molecules, and \((2\pm10)\) pH. The sensitivity of this micro- or nanoscope can be estimated as \(H = 10^{-4} (A \cdot m^2/\text{Hz})\) with SNR equal to 10°. The sensitivity of an advanced first-order biogradiometer is equal to 3fT/√Hz. The smallest resolvable change in magnetic moment detected by this system in the band 10 Hz is 1 fJ/T.

Keywords: biosignals, SuFET, nanoFET, biosusceptibility, nanoscope, operating conditions

1. Introduction – Different sensors and diagnostic systems

The sensor converts information or energy from the measurement quantity into another quantity, especially an electrical quantity. This “intermediate” signal may be processed in some way [1]. There are sensor and transducers for electrical quantities especially for current, voltage and magnetic quantities. A biosensor is a device that incorporates a biologically active layer as the recognition element and converts the physical parameters of the biological interaction into a measurable analytical signal [2]. Thanks to advances in nanotechnology, microelectromechanical systems, molecular diagnostics, and several other technologies, biosensors are now being developed to detect everything from the first chemical signature of cancer to the presence of anthrax.

The technology behind biosensors sounds perfectly simple. Typically, there is a probe on the tip of the sensor that makes contact with blood or tissue. Because of its particular electrical, metallic, and chemical composition, the sensor is able to "react" in a telltale manner when it comes into contact with the target. That reaction is then magnified, quantified, illuminated with various stains and dyes, and then matched against the ever-expanding libraries of gene sequences and protein structures of diseases [3].

There are a number of methods and devices for transducing different biosignals (BSs) into recordable or measurable information. The transfer of nerve impulses (NIs) is the main data flow, which carries sensory information to the brain and control signals from it and the spinal cord to the limbs. That is why detecting currents between neighbouring neurons and ionic currents in the nerve fibres is an important area of research. Electric-field control of physical properties is highly desirable from fundamental and technological viewpoints because it does not introduce any chemical or microscopic structural disorder in the pristine material. This is also the basis of FETs, in which accumulation, depletion, and inversion layers are formed at the interface [4]. Moreover, the complex view on BSs requires further stages of precise processing in order to decode the received or control information.

There are different kinds of transducers/sensors for picking up NIs: room-temperature and superconducting, external and implantable. Development of such devices is increasing the penetration into bioprocess while simultaneously simplifying the exploitation of the measuring systems in order to bring them closer to a wide range of applications. For this reason the magnetometer with a room-temperature pickup coil (PC) (without using the toroidal core) for detecting signals, which can clearly be detected in higher frequency range, was developed in order to simplify the superconducting quantum interference device (SQUID) system. The PC is set outside the cryostat and is connected to the input coil of the SQUID [5] or a channel of superconducting FET (SuFET) [6]. At frequencies higher than 20 Hz the noise decreases as 1/f function and then reaches a high flat noise at 2kHz as 200 fT/√Hz. On the other hand, implantable into nerve fiber transducers are evolving from the ordinary Si-chip microelectronics devices [7] into superconducting and nanodevices [8, 9].

The growing variety of biosensors can be grouped into two categories: implantable and external. Because external sensors employ widely used contact mechanisms like needles and lasers, this branch has advanced faster than implantables. For this reason a working interface between the living tissue of individual neurons and the inorganic compounds of silicon chips were developed. There also was provided the link between ionic channels of the neurons and semiconductor material in a way that neural electrical signals could be passed to the silicon chip. Once there, that signal can be recorded using the chip’s transistors. What’s more, the neurons can also be stimulated through the capacitors. This is what enables the two-way communications. The project tested the
device by stimulating the neurons and recording which ones fired using standard neuroscience techniques while tracking the signals coming from the chip [10].

The recent achievements in nanoelectronics can be regarded as a further step in the progress of BS transduction. They give us the possibility to create the most advanced and universal device on the basis of known micro systems. Such a sensor/transducer is suitable for picking up BSs- NIs, electrically active (ionised) molecules and the base-pair recognition event in DNA sequences- and transforming it into recognizable information in the form of electric voltage, or a concentration of organic or chemical substances. Moreover, this process can be executed in reverse. Substances and/or voltages influence BSs, whereby controlling or creating them (BSs). Also the new fiber-based optical microfluidic detectors are designed for nano-volume sample measurement. The silicon-glass microdetectors were fabricated by use of micromechanical techniques [11].

The known development of scanning-probe magnetic microscopes is based on circulating Josephson currents in a SQUID loop to produce microwave images with a spatial resolution of about 30 µm [12]. A SQUID-based microwave microscope has some distinct advantages over earlier microwave systems: the spatial resolution is limited only by the size of the SQUID loop. A mice biomagnetic measurement system using a dc SQUID magnetometer for a comparative magnetoencephalogram study of transgenic and wild mice has been developed [13]. A magnetometer with PC of 1.0 mm diameter and 0.7 mm lift-off distance to improve spatial resolution and magnetic field sensitivity was adopted. The magnetocardiogram of a wild mouse was measured as the initial application of the system. The bipolar peak positions in a magnetic field map separated by 5.6 mm were clearly detected with the maximum amplitude of 88 pT.

The advent of semiconductor devices with nanoscale dimensions creates the potential to integrate nanoelectronics and optoelectronic devices with a great variety of biological systems. Moreover, the advances in nanotechnology are opening the way to achieving direct electrical contact of nanoelectronic structures with electrically and electrochemically active subcellular structures-including ion channels, receptors, and transmembrane proteins such as bacteriorhodopsin. Direct electrical interfacing at the biomolecular level opens the possibility of monitoring and controlling critical biological functions and processes in unprecedented ways, and portends a vast array of possibilities such as new classes of prosthetic devices, medical monitoring devices, medical delivery systems, and patient monitoring systems, as well as other applications [14].

2. Sensors and diagnostic instruments: the input signals and flows

Electrical current may be measured by measuring the related magnetic field by means of the Hall effect or the Faraday effect into optoelectronic devices. The measurement as well as the reference value are often converted into quantities of either the same or different physical nature, before they are actually compared with each other [1]. Proceeding from the previously mentioned difficulties, including superconducting element of the sensor/transducer into an electric current could be the solution to the problem. Electronic or ionic currents in conductors or axons respectively, passing through the SuFET’s channel induce the output voltage on its gate [8].

Whereas the sensor element can deliver a weak signal, the transmitted signal should generally have a high signal level, and perhaps suitable values, in order to reach superior units undisturbed and to simplify the following calculations. Therefore, the sensor signal should be generally pre-processed. Thereby several important tasks could be realized, such as signal amplification, scaling, linearization, conversion, and conjunction with other components in a chain, parallel, or closed-loop structure [15].

As an electrical signal, the BS has two components: electrical potential or voltage and ionic currents. The first component is sufficiently developed and does not require penetration into the substances of BS propagation. The marketable progress in transducing of the second component began when the necessary instrumentation for measurement of micro and nano dimensions had been created.

The main informational flux from organs of the senses to motor nerves is transmitted through nerve fibres, which consist of a myelin shield with axons as a core. Recent research results suggest that such an arrangement is similar to a transmission line [16]. The nerve impulse in motor nerve of a frog is equal to 2 nA [17]. Synaptic currents between first order neighbouring neurons into in vivo or brain slice preparations have an order of 50 pA [18].

The nerve impulses passing through the fibre could be unambiguously defined by detecting the matching ionic current(s) or its superposition. Such a technique seems optimal because even precise voltage measurement could not give a current value according to the Ohm law. First of all, nerve fibre must be separated from a living organism for resistance of fibre measurement and, secondly, this resistance may vary in time. The implantable microelectrodes for neural applications are based on thin-film polymer foils with embedded microelectrodes for both recording and stimulation. Applications for these biomedical micro devices will include stem cell research, cancer cell characterization, drug discovery, treatments for neurological disorders, and neuroprosthetic devices [19].

A. The NIs propagation, measurement and simulation

The signal conducted along the axon is a transient. This difference in potential is the result of ionic gradients due to ionic concentration and is modiﬁed by the ionic ﬂow that produces ionic currents. These ionic currents give rise in turn to longitudinal currents closing local ionic current circuits that allow the regeneration of the membrane potential changes in a different region of the axon. This process is a true propagation instead of the conduction phenomenon occurring in wires.
Fig. 1. Longitudinal section of an axon showing a few lines of current flow.

The propagation of the NI in the cylindrical geometry of the axon and the consequences of the geometry of the axon on the current circulation across the membrane and along the axon is shown in Fig. 1 [20]. The generation of the action potential has been carried out in an axon with an axial wire.

Fig. 2. A nerve action potential propagates from proximal to distal (left to right in the figure). The wide and narrow arrows around the nerve represent the magnetic and electric field, respectively; the arrows on the nerve axis are equivalent dipole sources.

Fig. 3. The magnetic (solid lines) and electric (dashed lines) signals recorded from a frog sciatic nerve immersed in Ringer solution.

The magnetic fields were recorded with a SQUID magnetometer with a room-temperature PC [21]. At the closest PC-to-nerve separation of 15 mm, the nerve magnetic field could barely be detected (Fig. 2). The magnetic signal is produced predominantly by the total axial current $i_{ax}$ enclosed by the toroid while the voltage action potential in air expresses the charge on the membrane. The magnetic trace should then be proportional to the first derivative of the electric trace, as observed in Fig. 3.

Fig. 4. Test of the single-fiber model. Propagation of action potentials in an isolated normal fiber.

Development of such devices is increasing the penetration into bioprocess while simultaneously simplifying the exploitation of the measuring systems in order to bring them closer to the wide range of applications. Since the field and spatial resolution are highly diminished as the distance between the sample and the sensor increases, the key to this technique is to bring the sensor as close as possible to the sample. The cryogenic positioning system consists of three different main components and is very bulky. SQUID magnetometer systems do not provide the spatial resolution necessary to study the generation of the magnetic activity or injury currents at tissue and cellular scales.

Synchronous impulse transmission and the formation of "condensed" pulse states are found [22]. Electric impulses with a delay of 0.5 ms are presented to the system, and the numerical results show that, for increasing coupling, the impulses tend to adjust their speed and become synchronized (Fig. 4). Cell polarity is critical in various cellular processes ranging from cell migration to asymmetric cell division and axon and dendrite specification. Additionally, Par-3 directly associates and recruits the p75 neurotrophin receptor to the axon-glial junction, forming a complex necessary for myelination. Together, these results point to a critical role in the establishment of cell polarity for myelination [23].

B. Transducing and simulation of the neuron activity

Cultures of neurons can be grown on microelectrode arrays, so that their spike and burst activity can be monitored by a low noise multichannel integrated circuit for recording neuronal signals. These activity patterns are quite sensitive to changes in the environment, such as chemical exposure, and hence the cultures can be used as biosensors [24]. The signalling from the chip through a pair of neurons and back to the chip is thought to be the as a fundamental pathway of future neurochips (Fig. 5).
[25]. The activity of the fundamental element of such future hybrids was recorded by joining a silicon chip with an excitatory chemical synapse between a pair of identified neurons [26].

Modern neurophysiological experiments often require high-density multi-electrode readout systems for recording signals simultaneously from many neurons. Multi-channel readout systems are built mostly in hybrid technologies using off-the-shelf components. With such techniques readout systems with up to several tens of channels have been fabricated [27]. The development described in this paper is driven mainly by a project to readout signals from live retina tissue. Similar readout systems can be used in the study of slices of brain tissue and systems of cultured neurons.

In the experiments with retina, the amplitudes of extra cellular neuronal signals are typically in the range (50–500) µV with the frequency spectrum in the range (20÷2000) Hz (Fig. 6). Both the low signal amplitude and the low frequency range are challenging for an IC design. A low noise preamplifier and proper shaping of the frequency passband are required to optimise the signal-to-noise ratio. In most low frequency applications, high precision switched-capacitor filters play the dominant role.

The nervous system uses two basic types of formats for encoding information. The parameters of many sensory (and some premotor) signals are represented by the pattern of activity among an array of neurons each of which is optimally responsive to a different parameter value [28]. This type of code is commonly referred to as a place code. Motor commands, in contrast, use rate coding: the desired force of a muscle is specified as a monotonic function of the aggregate rate of discharge across all of its motor neurons. Generating movements based on sensory information often requires converting signals from a place code to a rate code.

Sensory maps or place codes are ubiquitous throughout the brain. The individual neurons that make up sensory maps respond maximally to different “preferred” values of the sensory parameter being encoded, and their responses decrease if the actual stimulus either exceeds or falls short of that particular favourite parameter value. Thus, only by examining the responses of a population of such receptors is it possible to deduce what the stimulus actually looks, sounds, feels, smells, or tastes like.

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recruiting additional motor neurons, or both (Fig. 7). The amplitude of the evoked movement initially varies with the number of pulses, then saturates. This behaviour is most similar to the summation-saturation model.

Fig. 8. Genosensor technology.
A - An atomic force microscope image of a 500-base-long (~250 nm) RecA nucleoprotein filament (black arrow) localized at a homologous sequence on λ-DNA scaffold molecule [34];
B - Expression levels of cell clusters in a microarray are proportional, over a fourfold range, to the amount of plasmid DNA [35].

The implantable electronic circuits for neural recording and stimulation in freely behaving animals that could provide prosthetic connections to replace or augment damaged pathways in the nervous system have been developing [29]. A Neurochip creates an artificial connection between two sites by using action potentials recorded on one electrode to trigger electrical stimuli delivered to another. The chip made the recordings and stimulations through two implanted electrodes, thus providing a constant connection between the two sites in the brain.

C. Analytical signal of DNA
Recent advances in automated DNA synthesis and the convenient site-specific labelling of synthetic oligonucleotides with suitable functional moieties, coupled with advances in microelectronics, have accelerated the development of biosensors for the analysis of specific gene sequences (Fig. 8). A DNA biosensor (or genosensor) employs an immobilised DNA as the recognition element [2]. Electrochemical DNA biosensors rely on the conversion of the base-pair recognition event into a useful electrical signal. Current DNA detection methods based on sequencing by synthesis rely on optical readouts; however, a direct electrical detection method for this technique is not available. Application of the label-free electrical detection method, charge perturbation detection, applied to sequencing by synthesis, could be used for direct electrical detection of enzymatically catalysed DNA synthesis by induced surface charge perturbation to detect any enzymatic DNA or RNA synthesis as well as other biochemical reactions based on similar principles [30].

D. Non-invasive biomedical instruments
SQUIDs have been a key factor in the development and commercialisation of ultrasensitive electric and magnetic measurement systems. In many cases, SQUID instrumentation offers the ability to make measurements where no other methodology is possible. Although SQUID electronics have the capability to operate well above 1 MHz, most applications tend to be at lower frequencies. Specific examples of input circuits and detection coil configuration for different applications and environments, along with expected performance, are described [31]. In particular, anticipated signal strength, magnetic field environment (applied field and external noise), and cryogenic requirements are discussed. Finally, a variety of applications with specific examples in the areas of electromagnetic, material property, non-destructive test and evaluation, and biomedical measurements are reviewed.

Since the field and spatial resolution are highly diminished as the distance between the sample and the sensor increases, the key to this technique is to bring the sensor, (held in at cryogenic temperatures), as close as possible to the sample. It has been shown that the best combination of spatial resolution and field sensitivity for a specific SQUID geometry occurs when the diameter of the pickup coil is approximately equal to the sample-to-sensor distance [32]. These systems do not provide the spatial resolution necessary to study the generation of the magnetic activity or injury currents at tissue and cellular scales. In excitable tissue extracellular potentials, transmembrane potentials, or action currents are interrelated. The extracellular potentials are typically recorded using microneedles arrays. However, the insertion of microneedles influences the measurement results and is impractical to achieve submillimeter (nano) special resolution. The approach to record the action currents using SQUID microscopy allows us to obtain more detailed information on the generation of the magnetocardiogram (MCG). High-resolution biomagnetic imaging provides insights that will improve existing mathematical models of biological tissue.

3. The components for superconducting nanosensors
CNT-based nanobiosensors may be used to detect DNA sequences in the body. These instruments detect a very specific piece of DNA that may be related to a particular disease. The use of nanotube-based sensors will avoid problems associated with the current much-larger implantable sensors, which can cause inflammation [33]. The devices can be administered transdermally. CNT chemical sensors for liquids can be used for blood analysis (for example, detecting sodium or finding pH value). Implantable sensors can be useful in health assessment. Because of their small size and less power consumption, they are highly suitable as implantable sensors. One way of compensating the loss of photoreceptors is by bypassing the destroyed photoreceptors and artificially stimulating the intact cells in the neighbourhood. Another possible area related to the application of CNTs that can be investigated is cochlear implants related to hearing problems.
Superconducting nanowires are unusual in that they newer show zero resistance, although resistance does exponentially upon cooling [34]. A new class of metallic devices based on DNA molecules is promising due to the self-assembly properties of DNA. As the resistance of the devices is controlled by the spatial profile of superconducting phase within the leads, there is the potential for applications. These include local magnetometry (as is widely done with conventional SQUID) and the imaging of phase profiles created by supercurrents – in essence a superconducting phase gradiometer.

Traditional materials have been pushed to their limits, which means that entirely new materials (such as high-kappa gate dielectrics and metal gate electrodes), and new device structures are required [35]. Entirely new device structures (such as nanowire or molecular devices) and computational paradigms will almost certainly be needed to improve performance. The development of new nanoscale electronic devices and materials places increasingly stringent requirements on metrology.

A. The Available FET’s Variants Applicable as SuFET

Application of the SuFET’s modifications such as CMOSuFET (low Tc) [36] and coplanar SuFET (high Tc) [37] broadens the range of requirements, which are being satisfied by the SuFET based transducer (SuFETTr). Low-dimensional semiconductor nanostructures and organic molecules, which offer unique possibilities such as extremely low power dissipation, quantum effects, surface sensitivity and low synthesis cost, could be the building blocks for next-generation electronics [38]. Moreover, an organic superconductivity of carbon molecules, known as bucky balls, which can act as superconductors at relatively warm temperatures, raises hopes for loss-free organic electronics and their practical applications in biosensors, including organic ones.

![Fig. 9. An organic SuFET device and its electrodes.](image)

High densities of electrons and holes have been induced by gate doping in a FET geometry (Fig. 9). At low temperatures, the material turns superconducting with a maximum transition temperature of 117 K in hole-doped C_{60}/CHBr [39]. The increased spacing between the C_{60} molecules increases the density of states, and the resulting increase of T_{c} is well documented in alkali metal-doped bulk samples (A_{x}C_{60}). The observation of gate-induced hole doping of C_{60} resulting in a T_{c} of 52 K suggests that significantly higher T_{c}'s could be anticipated in suitably “expanded” C_{60} crystals. Indeed, here is reported a raise in T_{c} to 117 K with such methods.

![Fig. 10. A - SAMFET structure. A highly doped Si substrate is used as the gate electrode, a thermally grown SiO layer acts as gate insulator, the gold source electrode is deposited by thermal evaporation, and the active semiconductor material is a two-component SAM of alkanedithiol mixed with 4,49-biphenyldithiol or 5,59-terthiophenedithiol. The drain contact is defined by shallow angle shadow evaporation of gold. The active region of the device is magnified; B - Transfer characteristics at room temperature of two SAMFETs (drain-source voltage of 20.5 V). The control corresponds to a “pure” 4,49-biphenyldithiol SAM, whereas the second one is based on a two-component SAM (4,49-biphenyldithiol to alkanedithiol ratio is 1:100). The inset shows the current at a gate and drain-source voltage of 20.5 V as a function of the mixture ratio.](image)

![Fig. 11. OFET of organic single crystals with bottom-contact configuration](image)
B. Organic FETs with a superconducting potential

Vertical self-assembled monolayer (SAM) FET (SAMFET) action, i.e. conductance modulation through a third electrode, in devices consisting of only several electrically active molecules has been reported [40]. The channel length of 10 to 20 Å is defined by the thickness of a SAM rather than lithography (Fig. 10). Moreover, the peak conductance at low temperatures is quantized in units of $2e^2/h$, indicating that the conductance of single molecules is modulated.

Organic FETs (OFETs) are of great interest for future electrolitic applications due to their flexibility. Up to now, a lot of fabrication processes or device configurations of OFETs have been reported [41]. Most of them were based on thin film technique such as vapour deposition. In general, the thin films for example, amorphous, poly crystalline, polymeric, and so on, consist of grains, and therefore, charge carriers behave as hopping conduction. This decreases the field-effect mobility of devices. Thus, annealing the organic thin films or surface preparations of substrate grows grains, and consequently these devices have few boundaries of grains between source and drain electrodes.

A quite simple and easy fabrication technique of OFETs with single crystals was proposed, and it demonstrated that the OFETs exhibit high mobility and good device characteristics (Fig. 11). The presented technique makes the fabrication of high performance OFETs easy and serves as a potential way to make organic integrated circuits. Also N-channel and ambipolar OFETs with a few tens of nanometer channel length were fabricated and characterized [42].

C. NanoFETs with Superconducting CNT Channel (nano SuFETs)

The first such devices were fabricated in 1998. The CNT was simply laid on the gold electrodes and was held by weak van der Waals forces. In addition to increasing the gate capacitance, it is essential that each CNT FET is gated independently by its own gate so that complex integrated circuits can be built [44]. The next generation of CNTFETs with top gates was fabricated by dispersing single-walled (SW) CNTs (SWCNTs) on an oxidized wafer (Fig. 14).

The intrinsic transconductance of CNTFETs in which CNTs were grown by chemical vapour deposition was measured at a drain voltage of -1V was 8.7 mS for a CNT
with a diameter of 1.5 nm (Fig. 15). Estimated intrinsic transconductance was 20 mS when parasitic resistance was taken into account. Measured and intrinsic transconductance per unit channel width were 5800 mS/mm and 13000 mS/mm, respectively.

This is considerably larger than those for state-of-the-art Si-MOSFETs. This transconductance can be improved drastically by decreasing parasitic resistance and the performance of CNT FETs will advance further by improving the CNT quality and the device structures [45].

The importance of being able to address nanoscale elements in arrays goes beyond the area of nanocomputing and will be critical to the realization of other integrated nanosystems such as chemical/biological sensors. A regular crossed-nanowire FET (cNW-FET) array (Fig. 16 A) that consists of n-input i and m-output j NWs, in which outputs are the active channels of FETs and the inputs function as gate electrodes that turn these output lines on and off [46]. Conductance versus applied NW input gate voltage data (Fig. 16 B) shows that each of the four cNW-FET elements could be turned off with gate voltage of 1V to 2 V, whereas the off-diagonal elements remained unaffected for the same input voltage. When NW are superconducting and, as a result, all cNW-FETs are functioning in SuFET mode, such output voltage V_out will show the changes in their conductance.

The realization of a self-assembled SW CNT FET operating at room temperature promotes such strategy as realistic for construction CNT-based electronics (Fig. 17). The assembly process was guided by the information encoded in DNA molecules and homologous genetic recombination [47].

Using a variant of nano-patterning, a self-assembling polymer could also create tiny, finger-shaped silicon protrusions sticking up from the underlying substrate [48]. These fingers would constitute the “channel” in a transistor through which electrons flow- but one in which electrons flow vertically instead of across a chip, as in
today's devices (Fig. 18). The gate to turn the transistor off and on could encircle the silicon finger. The geometry might prevent electrons from "tunneling," or leaking, through the channel when the transistor is in the off state, a constant threat when feature sizes become very small.

Fig. 18. Self-assembled nanocrystal FET. A layer of self-assembled silicon nanocrystals is inserted into an otherwise standard device.

The excellent device characteristics of SW CNT transistors in salty water also indicate that they may be ideal for biosensing applications. Since a SW CNT has dimensions comparable to typical biomolecules (e.g., DNA, whose width is approximately 2 nm), they should be capable of electrical sensing of single biomolecules [49]. A charged molecule near the SW CNT will act as an effective gate, changing the conductance of the tube. The large transconductances indicate that the signal from single molecules should be readily observable (Fig. 19).

4. Implantable Sensors: Design and Performance Data

In general, single sensor systems can only provide partial information on the state of the environment, while multisensor systems combine related data from multiple similar and/or different sensors. The goal of using multisensor systems is to provide synergetic effects that enhance the quality and availability of information about the state of the measurement environment [15].

A number of sensing organs of different physical quantities and environmental conditions are discovered in living beings. On the other hand, output signals from these organs—nerve impulses—can be picked up by state-to-the-art techniques. Further signal processing will define the connection between sensed quantity and output data of such biological sensors. As a result they could be applied in addition to the known artificial sensors for intelligent sensing.

Electrical properties of hybrid structures consisting of arrays of nanowire FETs integrated with the individual axons and dendrites of live mammalian neurons have been reported [53]. Arrays of nanowire-neuron junctions enable simultaneous measurement of the rate, amplitude, and shape of signals propagating along individual axons and dendrites. The configuration of nanowire-axon junctions in arrays, as both inputs and outputs, makes...

Fig. 19. Schematic of the electrolyte gate measurement. A water-gate voltage \( V_{\text{wg}} \) applied to a silver wire in the pipette is used to establish the electrochemical potential in the electrolyte relative to the device. For \(-0.9 \text{ } V < V_{\text{wg}} < 0.9 \text{ } V\), the leakage current between the water and the Au electrode/SWNT was negligible (less than 1 nA); the electrolyte then functions as a well-insulated liquid gate.

The ability to resolve separate sources within the brain or heart should be an important factor for all designs [51]. Different existing instruments can be compared by the dependence of spatial resolving power of the instrument on them. In designing a new system, given the physical constraints of the design, these methods may be used to optimise the system for resolving sources under the experimental conditions anticipated. The vector current sources are the usual equivalent current dipole sources that are the most frequently utilized source models for biomagnetic phenomenon.

The design parameters were the radius (or other geometrical constraints) of the array, the number of sensors, and the spacing of the sensors. Three basic gradiometers are possible [51]. With regards to gradiometer parameters, the type of gradiometer-planar, axial, or normal—and the base line of the gradiometer are important considerations. Finally, the dependence on the sources involves the relative orientation of the sources, the strength of the sources, the number of active sources, and the signal-to-noise ratios (SNR).

An axial first-order gradiometer is formed by coupling the input coil of a flexible high-temperature superconducting flux transformer inductively to a directly coupled SQUID magnetometer [52]. The tape of the transformer is bent such that the two outer pickup loops of the transformer are facing each other while perpendicular to the magnetometer plane resulting in a gradiometer baseline of 35 mm. A superconducting shield is mechanically adjusted to reduce the gradiometer response to uniform fields applied perpendicularly to both the magnetometer plane and the plane of the transformer pickup loops, by a factor of typically 7000.
possible controlled studies of partial to complete inhibition of signal propagation by both local electrical and chemical stimuli. In addition, nanowire-axon junction arrays were integrated and tested at a level of at least 50 "artificial synapses" per neuron.

Interfacing of nerve cells and field-effect transistors is determined by current flow along the electrical resistance of the cell-chip junction [54]. A spectral power density of the junction is (5\cdot10^{-11})W/Hz and can be interpreted as Nyquist noise of the cell-chip junction with a resistance of 3 M\Omega by measuring the fluctuations of extracellular voltage with a low-noise transistor. The thermal noise allows us to elucidate the properties of cell adhesion and it sets a thermodynamical limit for the signal-to-noise ratio of neuroelectronic interfacing.

A. Methods of picking up the biosignals: non-invasive and implantable, electro/magnetic- and biosensors

Voltage potentials of the living organism and its organs are measured by both implantable and external electric field probes of high sensitivity [55]. Information on organ activity is obtained by measuring biomagnetic signals. For such purposes a multi-channel high temperature superconducting interference (high T, SQUID) system for magnetocardiography (MCG) and magnetoencephalography (MEG) of humans, with high magnetic field resolution has been developed [56, 57].

The known amperometric techniques of biosignals involve the Renqvist bright realising method [17], and the second method of "biosensors typically rely on an enzyme system which catalytically converts electrochemically non-active analytes into products which can be oxidized or reduced at a working electrode which is main-tained at a specific potential with respect to a reference electrode" [58]. The same method is applicable also to potentiometric measurements "that can measure substrates, inhibitors or modulators of the enzyme". The Renqvist method requires extra stimulating of the isolated nerve fibre and the other method needs additional reagents and applied voltage. There are two feasible courses for impulse detection. The first one is measuring the magnetic flux around the fibres by PC [59]. The second one is to let the impulses pass through the electronic device with minimal impediments.

B. Devices for matching of biosignal to electronic element (circuit)

Creative integration of microchip technologies and nanostructures is feasible. By tuning the dielectrophoretic frequency within a microdevice, nanoparticles can be manipulated with the same precision as cells because a one-to-one correspondence exists between a given alternating current frequency and a nanoparticle interaction or biological event. Multiple biological events could be probed simultaneously provided that their corresponding frequencies are distinct. Combined with electroporation, electrophoresis also enables inclusion of molecular complexes inside the cells. Alternatively, functionalized nanoposts can be used to impale cells and relay information from the cell interior to neuroelectronic circuits. By merging the fields of microfluidics, electrophoresis, and cell biology, microchips are capable of creating tiny, mobile laboratories. The challenge for the future of designing a nano-interface in a microfluidic chip to probe a living cell lies in seamlessly integrating techniques into a robust and versatile, yet reliable, platform [60].

A planar FET can be configured as a sensor by modifying the gate oxide (without gate electrode) with molecular receptors or a selective membrane for the analyte of interest. Binding of a charged species then results in depletion or accumulation of carriers within the transistor structure. An attractive feature of such chemically sensitive FETs is that binding can be monitored by a direct change in conductance or related electrical property, although the sensitivity and potential for integration are limited. The so-called floating gate architecture combines a complementary metal oxide semiconductor (CMOS)-type n-channel FET with an independent sensing area for recording extracellular signals from electrogenic cells was presented [61]. This concept allows the transistor and sensing area to be optimised separately. The noise level of the devices was smaller than of comparable non-metallised gate FETs. The potential of NW nanosensors with direct, highly sensitive real-time detection of chemical and biological species in aqueous solution has been demonstrated [9]. A silicon NW (SiNW) solid state FET, whose conductance is modulated by an applied gate, is transformed into a pH nano-sensor (Fig. 20) by modifying the silicon oxide surface with 3-aminopropyltriethoxysilane (APTES).

The report shows how the scientists fabricate FETs from CNT with the precise electrical properties and any variable band-gap desired. In parallel studies of CNT, researchers have been working to improve the electrical characteristics of individual nanotube transistors [62].

C. Design of a SuFET based either implantable or non-invasive transducer of the BSs

The method of combining the bioelectric nature of NIs and synaptic currents between neighbouring neurons with body-temperature PC and zero resistance input of the SuFET device in order to obtain most advantageous biosensor/transducer was recently advanced [8].
The SuFET is used as a zero-resistance ammeter, which converts drain currents into gate voltages [6].

Transducing the vertical magnetic field from the BSs by the PC that is wrapped around the nerve fibre or DNA sequence is executed similarly to Fig. 2, but the PC is in nano-dimension [63]. By using the said superconducting magnetometer with a room-temperature PC (SIM) [6] it is possible to create the implantable transducer [8].

D. New nano-microscope and biosusceptometer

There was described the eddy current microscopy of thin conducting samples using a SQUID-based magnetic flux microscope in which the sample and SQUID are cooled in liquid nitrogen [64]. If the technique could be extended to the imaging of room temperature samples, it could be used to diagnostic purposes. To achieve a high spacial resolution and flux resolution, this microscope uses a small SQUID directly as a magnetic sensor, rather than a superconducting pickup loop coupled to SQUID. A time-varying field not only induces local eddy currents but also produces currents, which circulate around structures with closed loops. The ability to detect small regions of nonmagnetic conducting materials, and distinguish them from ferromagnets, superconductors, and paramagnets at the microscopic level, gives the magnetic flux microscope broad capabilities in materials analysis. For such testing to be of practical use, it will be necessary to develop systems, which allow the microscopic magnetic imaging of room temperature samples.

In order to realize a short distance between the cooled SQUID and the room temperature sample, a microscope-type SQUID has been developed [65]. In this system, the diameter of the sample is 5 mm. Corresponding to this sample size, a directly coupled thin film gradiometer with two 5×5 mm² pickup coils is used, where the gradiometer is chosen to reduce the spatially uniform external noise. The distance between the sample and the SQUID is approximately 1.5 mm, which is short enough for a sample size of 5 mm. The SQUID and the sample are surrounded with three layers of a permalloy magnetic shield in order to reduce the disturbance from the environmental magnetic noise.

A low temperature magnetic susceptibility was constructed by combining an Oxford Instruments Heliox probe sorption pumped 3He cryostat with a Quantum Design model 5000 DC SQUID sensor, giving an effective temperature range from 300 mK to 4.2 K [66]. The smallest resolvable change in magnetic moment detected by this system was found to be Delta~10⁻¹² J·T⁻¹.

1) In vivo micro- and nanoscope for scanning and sounding

A new design for scanning or sounding micro- or nanoscopes that combines a simple mechanical arrangement with a miniature SIM's PC. The microwave imaging process (Fig. 21) is shown on a prototype sample: a normal conducting ring of self-inductance L, and resistance R, The quantity L, is the PC self-inductance and MI, represents the magnetic coupling of the PC through a mutual inductance M to an external circuit carrying a current Ii in a tissue [67]. If \( \omega_r \gg R/L, \) then R, may be ignored and \( I(\omega_r) = I_r(\omega_r)M/L, \)
conducting materials, gives the magnetic flux nano-
microscope broad capabilities in material analysis. Since
the developed system allows the micro- and nanoscopic
of room and tissue temperature samples, such testing will
be of practical use for clinical diagnostic.

The sensitivity of this instrument \( H_s \) can be estimated
by considering the noise source \( I_{ir} \) of SuFET according to
Ref. [6]. The digital value of \( H_s \) for PC with the diameter
of 0.1 \( \mu \)m and inductance 1 \( \mu \)H, and \( I_{ir}=10^{-11} (A \cdot Hz) \)
is equal to: \( H_s=I_{ir}L_{ir}S_{ir}=10^5 (A \cdot m^2/Hz) \). This means
that with a SIM exciting signal \( H_{ex}=H_s=V_{simp}/I_{ir}L_{ir} \)
and \( I_{ir}=1 \), and is equal to: . This means
the noise-limited sensitivity was estimated from
measurements on a niobium foil of known mass to be
10 \( -14 \)Am Hz.

A technique had been previously developed, based on
magnetic field measurements using a SQUID sensor, to
localize in three dimensions steel needles lost in the
human body. In all six cases that were treated until now,
the technique allowed easy surgical localization of the
needles with high accuracy. Despite the importance of
needle localization, the most prevalent foreign body in
the modern society is the firearm projectile (bullet),
that creates a corresponding output voltage \( U_{out} \). PC has
been handling in any direction of the subject's irregular
form by a biochemically inert rod or manipulator 7.

For the biological tissues the magnetization \( M(r) \) in the
point \( r \) is defined by the value of magnetic
susceptibility \( \chi(r) \) as:

\[
M(r) = \chi(r) \cdot H(r)
\]

(4)

where \( H(r) \) is defined by the supplied current \( I_{ir} \) and
the transforming factor of the coils \( K_{ii} \) as: \( H(r) = K_{ii} \cdot I_{ir} \).
\( H_{lin} = U_{in} K_{ii} / R_z \). The value of \( M(r) \) with the
frequency \( f \) and energy resolution of SuFET \( E_p \) will be
defined as [6]:

\[
M(r) = \frac{e I_{ir} \mu_0 S_{eq} E_p \omega r f}{\pi \mu_0 S_{eq} E_p \omega r f}
\]

(5)

where

- \( \omega \) - the impedance of PC;
- \( \mu_0 \) - the permeability of free space, \( \mu_0 = 4 \cdot 10^{-7} \) henry/meter;
- \( E_p = 10^{-9} (J/Hz) \);
- \( S_{eq} \) - equivalent area of PC.

From Eq. (4) and Eq. (5):

\[
\chi(r) = \frac{e I_{ir} \mu_0 S_{eq} E_p \omega r f}{\pi \mu_0 S_{eq} E_p \omega r f}
\]

(6)

If we define \( K_{e} = e I_{ir} \mu_0 S_{eq} E_p \omega r f / \pi \mu_0 S_{eq} E_p \omega r f \)
as a factor of biosusceptibility meter, the value of \( \chi(r) \)
will be calculated simple:

\[
\chi(r) = K_{e} \cdot V_{GS} / U_{in}
\]

(7)
The organic or nano SuFET device shown, for example in Fig. 17, is connected with the source of BSs through the said micro or nano contacts according to Fig. 23. In such case the current of i, passing through the SuFET's...
channel arouses $V_{out}$ on the gate with some constant bias voltages $V_{bias}$, $V_{g}$ (Fig. 24). Transfer function (TF) of the device defined by its numerical order as follows:

$$K_{DNA} = \frac{V_{out}}{i} \Rightarrow \left[ 10 \right]_{0.5 \mu A} = 2 \cdot 10^3 \left[ \frac{B}{A} \right] \quad (8)$$

As the channel of the majority of cited FETs is not superconducting in the present stage of development, it is possible to define the sensitivity threshold (ST) by the channel’s resistance $R$:

$$U_{noise}^{DNA} = \sqrt{4kTR} \quad (9)$$

where $k$ - the Boltzmann constant, $T$ - absolute temperature of FET’s channel.

The noise voltage of SuFETs is obtained analogous to the Johnson noise value for the resistive circuit as [6]:

$$\left( E_n \right)^2 = 4kTR_{SuFET} \frac{g_{in}}{g_{in}} \quad \text{and} \quad g_{in} = I_c/V_c \quad (10)$$

where $g_{noise}$ is the ratio of the kinetic energy of the JJ link to thermal energy and $V_c$ - characteristic voltage. Ratio $g_{noise}$ is proportional to the normalized gate charge $g_{in}/C$ ($g_{in}$ is the total gate charge reflected in the channel).

The peak currents range of BS from 5 to 10 $\mu A$ [21] give a maximal output voltage $V_{out}$ on abscissa axis -5 to 5 V also with the necessity of some its reducing it slightly by changing $V_{bias}$ (bias) of the FET’s channel. A current, which elicits an action potential in the neuron, is 0.6 nA [25] and will stimulate $V_{out}$ of the transducer equal to 12 mV.

**B. Parallel Connection of BSs to SuFETTr**

An example of connecting the axons of nerve fibre to multi-input cNW-FET [46] array is shown in Fig. 25. The output voltage of the transducer according to Fig. 16.B lies in the range of 1 V (linear part of the curves).

![Fig. 25. Schematic of SuFETTr in the parallel connection.](image)

The operational pattern of the transducer for one input BS into cNW-FET is shown in Fig. 26. The input BS changes conductance of the FET’s channel which influences output voltage $V_{out}$.

TF of this device will be similar to the previous one:

$$K_{cNW} = \frac{V_{out}}{V_{sup}Q + i} \Rightarrow \left[ \frac{1B}{10^{-5}A} \right] = 10^7 \left[ \frac{B}{A} \right] \quad (11)$$

Similarly the value of $V_{out}$ can be obtained for organic, chemical and DNA SuFETTrs.

The carrier density thus obtained is a function of the distance $x$ from the interface (Fig. 11). The large carrier density at $V_{bias}=0$ V in small $x$ region is due to the charge transfer from the Au electrodes to the pentacene molecules [4]. Carrier density $n(x)$ is a function of the distance ($x$) from the interface at various gate voltages ($V_{g}$). According to the Eq.(4) deviation of carrier density $n(x)$ in the range of maximal $V_{out}= \pm 15$ V is equal to $(7+0) \cdot 10^7/cm^2$.

Chemical SuFETTr converts the changes in pH through $Q$ of the channel (similarly to pH nanosensor Fig. 20.A [9]) also into output signal $V_{out}$. In the scale of $Q$ from 10 to 400 nS a pH (2 to 10) is transformed into variations of $V_{out}$ from 0 to 1 V (Fig. 26).

The potential applied during immobilisation of thiol-modified DNA lies in a different potential window from -0.7 to +0.7 V [2]. The corresponding current $i=V_{in}/Q$, to this voltage which is put into the FET channel, according to Fig. 26 vary from 0.4 to 1.8 nA.

Clean gold monolayers are stable in the potential range from -400 to +1400 mV in dilute sulfuric acid solutions, thus allowing electrochemical applications. The careful selection of the terminal functionalities of the monolayers and the proper surface chemistry allows a tremendous flexibility in biosensor design.

ST of the biosensor/transducer in the first approximation (generally) depends on the conductance $Q$ of the superconducting channel. An absolute sensitivity of the transducer with $n$ parallel inputs in the said frequency range derives also from the Johnson noise:

$$U_{noise}^{cNW} = n \sqrt{4kT/|Q|} \quad (12)$$

which for cNW-FET varies in the range 50 to 150 nS [46] and gives the order of ST $10^4 V/\sqrt{Hz}$. Also for the noise voltage of parallel SuFETs based transducer is:

$$\left( E_n \right)^2 = 4nkT_{SuFET} \frac{g_{noise}}{g_{in}} \quad (13)$$

**7. Results**

Application variety of the novel superconducting, organic and CNT FETs allows us to design transducers of BSs (electronic, nerve, DNA, etc.) that transduce them into different quantities, including electric voltage, density of chemical and biomolecules. On the other hand, the said BSs can be controlled by the applied electrical signals, or bio and chemical mediums.

The described SuFETTrs designed on the basis of organic and nano SuFETs are suitable for describing the wide range of BS dynamical parameters (see Table 1).
Following the columns of the table, it should be noticeable, that serial connection of the external PCs allows us to gain some integrated signal, i.e., the whole sensing or control electronic or NI, which spreads along the number of axons of the nerve fibre; the amount of ions passing through the PCs and the generalized BS passing through one or both spirals of DNA. When SuFET channel(s) of are implanted into the tissue or process we can acquire more precise data about the frequency distribution of NIs, volume distribution of ionized molecules and detecting activity of individual nucleoteds.

Table 1. Dependence of the received BS parameters on the mode of SuFETTr’s functioning.

<table>
<thead>
<tr>
<th>Mode</th>
<th>Serial</th>
<th>Parallel</th>
</tr>
</thead>
<tbody>
<tr>
<td>Medium</td>
<td>external</td>
<td>implantable</td>
</tr>
<tr>
<td>Current</td>
<td>$i=\int \text{cont. or sens. imp.}$</td>
<td>$i=\int f_1(x_1)f_2(x_2)\cdots f_n(x_n)$</td>
</tr>
<tr>
<td>Molecules</td>
<td>$\Sigma \text{BSs} \rightarrow \text{bio}$ and chem. molec.</td>
<td>variation of BSs $\rightarrow$ concentr. of molec.</td>
</tr>
<tr>
<td>DNA</td>
<td>propagation of BS along DNA's spirals</td>
<td>decoding the BSs of nucleotided recognition</td>
</tr>
</tbody>
</table>

The described biosusceptometer or nano-microscope designed on the basis of SuFETTrs is suitable for investigating both the structure of organic objects and their comparing analysis (see Table 2). Following the strings of the table, it should be noticeable, that investigations of biological surfaces are performing according to the surface integrals for a biosusceptometer and nano-microscope modulus respectively. The surface gradients are acquiring by finding of the difference between the respective values of $I_1$ or $I_\text{c}$. The same is applying to the investigations of biological volumes $V_1$ and $V_\text{c}$ as the double and triple integrals respectively. The next two strings are explaining the bounds on the possible spreading of the said method.

Table 2. Dependence of the received structure parameters on the mode of functioning- a biosusceptometer or nano-microscope.

<table>
<thead>
<tr>
<th>Medium</th>
<th>Mode</th>
<th>biosusceptometer modulus</th>
<th>nano-microscope modulus</th>
<th>biosusceptometer gradient</th>
<th>nano-microscope gradient</th>
</tr>
</thead>
<tbody>
<tr>
<td>Surface</td>
<td>Investigation of sheath (envelopes) of organs</td>
<td>$I_1=\int f(x, y, z)dx$</td>
<td>$I_2=\int f(x, y)dy$</td>
<td>$\Delta I=I_1-I_2$</td>
<td>$\Delta I=I_1-I_2$</td>
</tr>
<tr>
<td>Volume</td>
<td>Investigation of the inside structure of the organs and tissues</td>
<td>$V_1=\int f(x, y, z)dv$</td>
<td>$V_2=\int f(x, y, z)dv$</td>
<td>$\Delta V=V_1-V_2$</td>
<td>$\Delta V=V_1-V_2$</td>
</tr>
<tr>
<td>Structule level</td>
<td>Differential investigation the twin (pair) organs or tissues</td>
<td>Comparing investigation of the twin (pair) organs or tissues</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Object (body) level</td>
<td>Investigation of homogeneous organs or tissues</td>
<td>Investigation of inhomogeneous organs or tissues</td>
<td>Comparing investigation of the homogeneous organ's or tissue's areas</td>
<td>Differential investigation the inhomogeneous twin (pair) organs or tissues</td>
<td></td>
</tr>
</tbody>
</table>

Exploitation of the parallel input to SuFETTr allows determination of space and time dynamics of BSs in the nerve fibre and DNA spiral(s) and also the amplification of output signal $U_\text{c}$ by multiplying the concentration of molecules according to a number of input BSs. After the implantation of parallel SuFET(s), the averaging or summation of this dynamic among the whole neural network, nerve fibre or DNA spiral(s) is possible.

8. Conclusions

The invented biotransducer has the following fundamental improvements upon existing ones:

a) the sign of the output voltage permits the determination of the direction of the input bioflow passing through a single SuFET device;

b) situating the reference electrode outside the living organism makes precise measurement possible;

c) the capability to regulate the proportion of axons, neurons or flows that are being investigated to the untouched ones- either the whole cross section of the fibre or flow, or any part of them;

d) the possibility to substitute the SuFET device or to adjust its ratings to comply with the conditions of the measurement process without repeatedly destroying nerve fibre or flow vessel;

e) the transducer could create conversion in both directions, respectively in passive and active modes;

f) the combination of biocompatibility and tissue equivalence in both the diamond and protein-based (organic) FETs makes them naturally fit for implantation;
g) the possibility to compose a converging device by changing the instrument - from a current sensor to biosusceptometer, from a flowmeter to nanomicroscope, etc. - by switching-over (modifying) of SuFET’s working mode or transferring from the natural ambient conditions to applying of the exciting magnetic field.

The reviewed variety of FETs shows the varying extent of readiness for them to be exploited them in SuFETTr of BSs. The most appropriate for such an application are the ordinary solid-state SuFET modifications and novel CNT based SuFETs. The organic SuFETs are not amply developed, but this work is being carried out in a number of directions. At the same time, the PCs, which are necessary for the external sensor with respect to the transducing medium (solid-state conductor, nerve fibre, flow of ions and DNA spiral), and corresponding low-ohmic wire traces for connecting PCs to the FET’s channel are sufficiently developed, even at nano dimensions.

The preliminary calculations confirm the possibility of broadening the SuFETTr’s action from magnetic field to the biochemical medium of BSs. The main parameters of such BSs can be gained by applying the arrangement of the SuFETTr(s) to the whole measurement system. Two directions of SuFETTr function enable decoding of the BS by comparing the result of its action on some process or organ with an action on them of the simulated electrical or biochemical signal after their reverse transducing through the SuFETTr(s). Furthermore, this decoded signal will provide a basis for creating feedback and feedforward loops in the measuring system for more precise and complete influence on the biochemical process.

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